Iterative Reconstruction Methods for High-Throughput PET Scanners

James Hamill and Thomas Bruckbauer
CTI PET Systems, Inc., 810 Innovation Drive, Knoxville, TN 37932 USA
e mail: hamill@cti-pet.com

Abstract. **Iterative reconstructions for clinical PET must run fast. We describe a clinical processing method based on sinogram rebinning, Fourier rebinning for the 3D to 2D data reduction, and iterative reconstruction using the attenuation-weighted OSEM method with a projector based on a gaussian pixel model. When this approach is used, multi-bed clinical oncology scans can be ready for diagnosis within minutes.**

Introduction

As PET becomes used increasingly in the clinic, the need for high throughput in whole-body oncology is being met in several ways. First, PET scanners with septa to block radiation on oblique lines of response (two-dimensional, or 2D, scanners) are being replaced by scanners with removable septa (3D capable scanners), and more recently by scanners with no septa (fully 3D scanners) to make use of the increased sensitivity in 3D mode. Second, detectors with a higher intrinsic count rate capability are being developed to allow more patients to be scanned in one day, for example, scanners with LSO scintillators instead of BGO. Third, algorithms are being developed to improve image quality. The most promising of these algorithms are iterative reconstruction methods.

The iterative approach to image reconstruction is computationally intensive. One could hope that the arrival of faster computers every few months will allow the use of increasingly more sophisticated algorithms, yet it has been observed [1] that, over the last two decades, the number of lines of response in PET scanners has grown at a rate that outpaces Moore's Law which describes the density with which foundries are able to pack transistors onto a microchip. Processing algorithms must also be coded to run very fast.

In this paper, we describe an approach to generating clinical images with the new generation of 3D PET scanners. The approach is the same one described in [2] but applied to 3D. Sinogram rebinning is an important feature of our approach, in which we reduce the size of the input data sets early in the reconstruction procedure.

Methods

Images of one position of the patient bed are derived from the 3D emission sinogram, a set of normalization coefficients, and attenuation correction factors (ACF). The sinograms represent an estimate of true coincidences, since random events are subtracted from the sinograms during acquisition. The ACF's may be taken directly from a transmission measurement, or they may result from a segmentation procedure.

To prepare the sinograms for reconstruction, we first multiply them by normalization coefficients. [3] Next, each plane in each oblique segment of the high-resolution sinogram is rebinned radially and angularly. Clinical users often use 128 sinogram bins and 128 angular ones. After rebinning, the sinogram contains all quanta in the normalized sinogram, but the arc distortion has been removed and the sinogram bins represent equally spaced, parallel lines of response. The rebinning algorithm is indicated in Figure 1, where: "source" coordinates are those of the normalized sinogram with arc distortion; "destination" coordinates are those of the rebinned sinogram; and left and right edges are determined by the coordinate transformation which changes the sinogram bin size and describes the arc distortion. The algorithm integrates the interpolation function (which is a step function in the case shown) over the needed range of source coordinate positions, thereby preserving the counts.

The transformation from 3D to 2D sinogram is performed with the Fourier rebinning method. [4] Before Fourier rebinning is performed, we make the sinograms more consistent by applying the segment-0 attenuation correction to all sinogram segments. After Fourier rebinning, we form an estimate of a 2D sinogram of scattered radiation, using Watson's method. [5] To create the inputs required by this method, the 2D emission sinogram is reconstructed and smoothed to create a low-resolution image which is known to be corrupted by scatter; and the logarithm of the ACF is reconstructed to obtain an image of attenuation coefficients. Both of these reconstruction steps use a filtered backprojection technique which is implemented in frequency space. Next, the attenua-

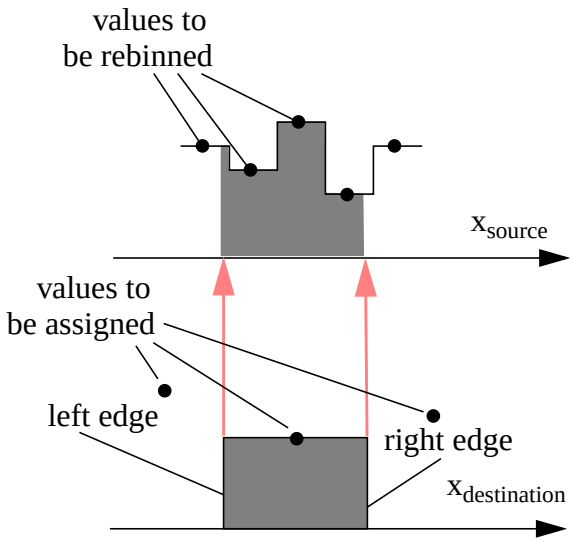

FIGURE 1. Rebinning method

tion coefficients are removed from the emission sinogram, and the scatter estimate is subtracted from it.

The reconstruction itself, that is, the inversion of the Radon transform, is performed with the attenuation-weighted OSEM method (AW-OSEM). [6] For the forward projector and the matched backprojector, we use a gaussian voxel model. [7] Negative sinogram values are set to zero before the iterative reconstruction begins. Our initial image estimate has the value 1.0 in all voxels. Clinical users often select the option to use eight subsets and to stop after two iterations. After reconstruction, the images are filtered transaxially and axially.

The convergence properties of this algorithm were presented in [2]. In this paper we present new clinical results from a fully 3D LSO scanner. Although it is not the thrust of this work, we compare AW-OSEM images to those from another, older OSEM reconstruction method, [8] in which attenuation factors are applied to sinograms before reconstruction. We call the old method UW- OSEM, where UW denotes unweighted.

Clinical Performance and Patient Images

This approach is designed to run fast. In a typical clinical acquistion, images are available about 3 min after all patient data is acquired in fully interleaved mode when the reconstruction is paralleled to the acquisition. In the case discussed in this article, the images were available when the patient left the scanning room. An analysis of the time involved indicates that about 45 seconds are used to read data over the network from the acquisition computer; 50 seconds to set up the 3D normalization matrix; about 20 seconds for the scatter correction; and approximately 20 seconds for every iteration on a typical computer configuration (Sun Ultra-60 running at 450 MHz).

The comparison of the UW-OSEM and AW-OSEM algorithms is shown in Figure 2. Clinicians normally prefer the results from the new, faster, attenuation weighted algorithm.

Image quality is demonstrated in Figure 3. The sinograms for this scan were acquired in 3D on the ECAT Accel, a new generation of LSO based PET scanners, in 45 min (9 bed positions, 3 min in emission scan time and 2 min transmission scan time each).

Conclusions and Future Work

We have realized an acceptable and, we think, high level of image quality with iterative reconstructions that run in clinically realistic times. The keys to doing the calculation fast enough are: sinogram rebinning and Fourier rebinning as techniques for reducing the number of sinogram bins; and use of the attenuation-weighted OSEM algorithm. The result is a procedure that allows routine 3D reconstruction of clinical images of as many patient scans as can be acquired during the day.

This model does not treat the background of random events accurately. Among the techniques to be explored in the future is NEC scaling. [9]

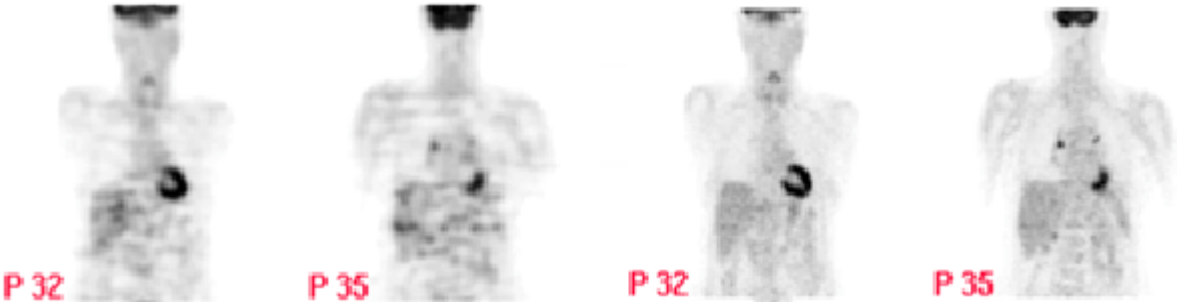

FIGURE 2. The left side shows two coronal slices through a typical data set reconstructed with the previous version of the iterative reconstruction using pre-corrected sinograms and 1 iteration, 30 subsets, and a 10 mm gaussian filter. On the right side, the same data set is reconstructed with the new algorithm and 2 iterations and 8 subsets. The parameters were chosen to generate images preferred by the clinicians for each algorithm

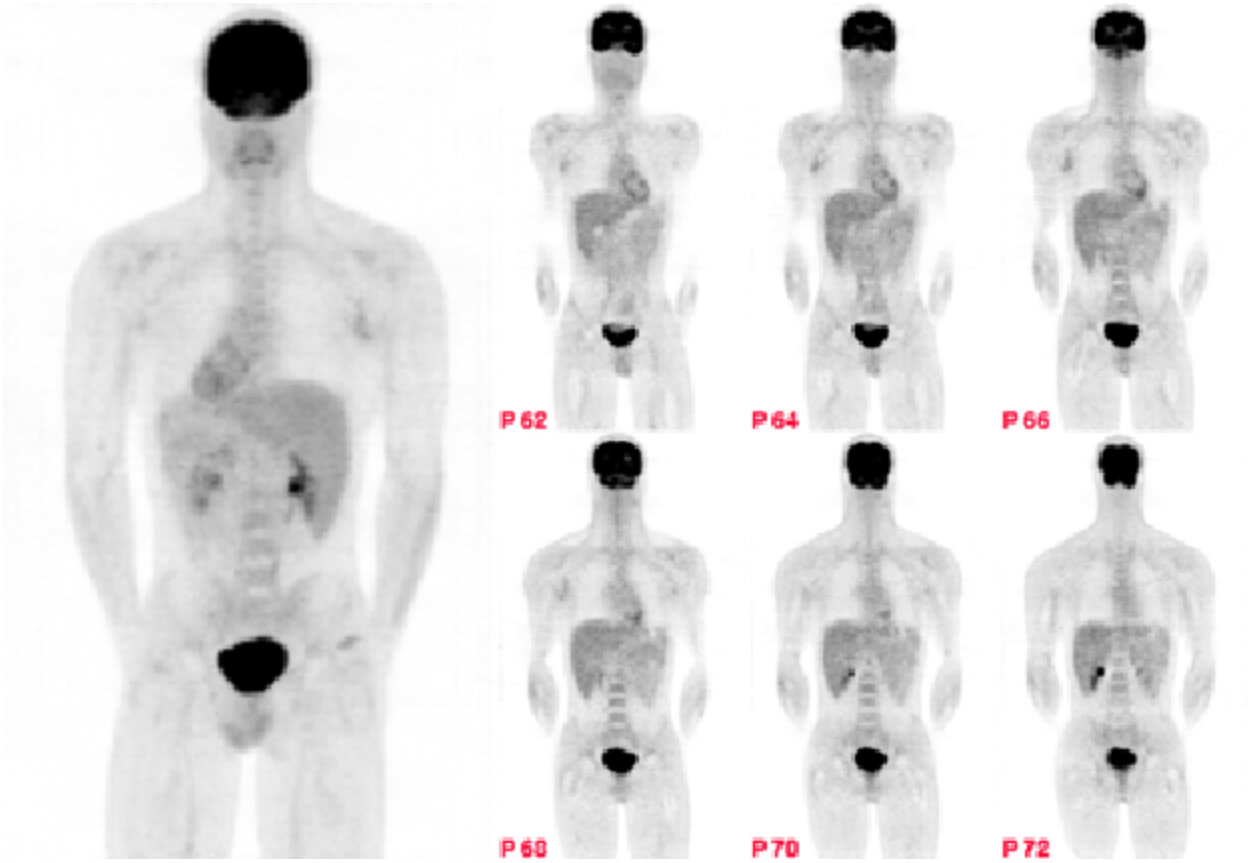

FIGURE 3. Image of a 25 year old, 71 kg, male melanoma patient 45 min post injection of 16 mCi FDG, 9 bed positions acquired for 3 min emission and 2 min transmission each.

The gaussian voxel model should be well suited to fully iterative 3D reconstruction. Schmand showed that the line spread function of high-resolution LSO and GSO brain scanners is well modeled as a gaussian. [10] See Figure 4. We are looking for ways to extend our algorithm to that reconstruction problem.

Acknowledgment

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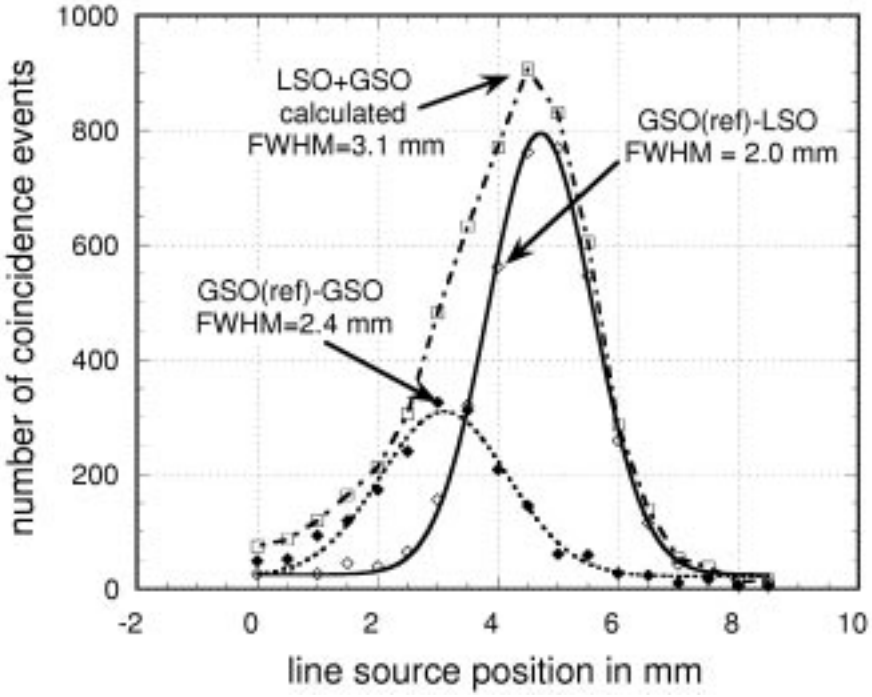

FIGURE 4. Line spread function for 45-degree incidence in the 2-mm crystals of a high resolution brain tomograph with 2-mm LSO and GSO crystals. From [10]

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